

## Short biography:

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### José Carlos Pimenta Claro

#### Education:

- Degree in Mechanical Engineering; and
- PhD. in Mechanical Design, with a specialization in Tribology, by the Universities of Minho, Portugal, and University of Leeds, UK

#### Positions:

- Associated Professor of the Mechanical Engineering Department, University of Minho, Portugal
- Senior researcher of the Mechanical and Materials Technologies (CT2M) research centre, of the Foundation for Science and Technology, Portugal

#### Main research fields:

- tribology - thermo-hydrodynamic lubrication
- multibody - simulation of human joints biomechanics
- crash analysis - vehicle and pedestrian dynamic modeling

#### Scientific work:

- Researcher in 3 national, 1 MIT/EDAM, 1 EC-7th.FP funding projects; and
- Co-author of 1 book, 11 papers, 19 communications and 1 patent (in the last 5 years)

#### Member of the:

- scientific committee (and co-founder) of the ***IberTrib - Iberian Congress on Tribology***
- board for projects evaluation of the ***Georgia Science Foundation***
- reviewers' board of ***Tribology International***, ***Mechanical Behavior of Biomedical Materials*** and ***International Journal of Computational Methods***
- SAE International, former ***Society of Automotive Engineers***

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### José Luís Martins Alves

Assistant Professor of the Department of Mechanical Engineering of the School of Engineering of the University of Minho, Portugal, received his PhD degree from the same University in 2003.

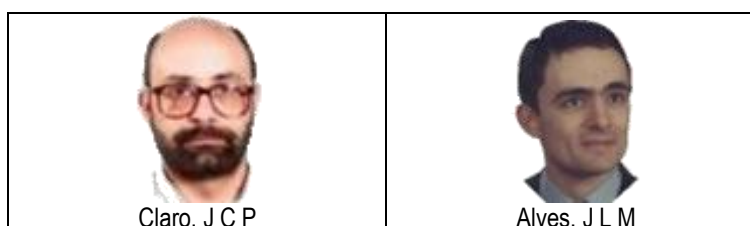
In 2006 was awarded with the ESAFORM Scientific Prize.

Developed his research activities on several international institutes, namely, CEMUC (Portugal), LPMTM-CNRS (France), RIKEN Institute (Japan) and, more recently, as Adjunct Associate professor at REEF/University of Florida (USA).

Current research interests are in computational mechanics, computational plasticity and model materials, along with biomechanics, in the perspective of modeling either the musculoskeletal system or the hyper-visco-poro-elastic behavior of soft-tissues.

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## Photos:



Session:

## Symposium I - What is known about the disc space

Title:

### Disc mechanical characteristics – Construction of a finite element mathematical model, first results

Authors:

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#### Abstract

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##### 1. Introduction

Computational mechanics is an invaluable tool to analyze biomechanical systems, either in healthy or degenerative conditions, and to improve our understanding on the events that can trigger trauma or diseases, to design new medical devices to restore working conditions, or even to point out treatment techniques.

Numerical methods in general, and the Finite Elements Analysis (FEA) in particular, if properly built and used, can allow an inside view, a rigorous analysis and a qualitative study of any assumption, frequently too much difficult or even impossible to achieve with any *in-vivo* or *in-vitro* experimental technique.

An Intervertebral Disc (IVD) is a functionally-oriented construction of several soft tissues, supporting a wide range of dynamic and static loads that generate complex stress fields, which experimental study and understanding of its biomechanical behavior is of an enormous complexity.

On the one hand, human's *in-vivo* study is almost impossible – due to the high degree of uncertainty in applied loads, geometric variability of individuals, complex surrounding musculoskeletal interactions, the role played by electro-chemical phenomena like osmolarity, etc – and *post-mortem* studies hardly provides accurate information to allow a clear and precise characterization and transposition to *in-vivo* biomechanics. On the other hand, due to that intrinsic complexity of the IVD, an accurate biomechanical model cannot easily be achieved. It is rather a step-by-step task where, although there are still many open questions, an important effort is being done to bring to the FEA the multi-physics behavior, and the complex interactions between them, in order to accurately model the IVD's constitutive performance.

This work is focused in the most relevant issues and phenomena that shall be taken into account in the development of an accurate biomechanical FEA model of the IVD, either in healthy or degenerated states.

##### 2. Challenges

###### 2.1 Geometric reconstruction

In general, a FEA mesh can be created from a solid voxel-based model, obtained through a 3D reconstruction of a set of 2D medical images (eventually using different techniques such as X-Ray, CT, MRI, etc.) or built up by algorithms using standard anatomical data.

Despite a 3D model based on specific data, significant 'average' geometrical differences can often be found. These discrepancies, usually imputed to the factual anatomical variability – or, less often but as well probable, due to the different imaging techniques used – may also reveal a lack of robustness, or simply difficulties on recognition and reconstruction of the 3D entities.

In one hand, the 2D source images' level of contrast (i.e. the range of the 'scale of grays') and resolution (i.e. the pixels size and slicing thickness), among other parameters (inherent to each imaging technique and equipment), may play a key role on the dimensional accuracy. On the other hand, soft tissues of the IVD – such as the nucleus pulposus (NP) and the annulus fibrosus (AF) – have too similar densities, making image segmentation a very user-dependent task.

Finally, in order to turn the process into an automatic or at least semi-automatic task, different segmentation algorithms can be employed – using intensity, contour/edge and regions detection, clustering, etc. – whose interpretation and manipulation of each pixel, their neighbors and boundaries definition can lead to substantial different results.

###### 2.2 Geometrical standardization

An alternative to the use of 'custom tailored' geometrical entities and models, especially when generic results are pursued, is to rely upon standardized anatomical data. However, in spite of a considerable effort being done in global terms (HFES, ANSI, ASTM, ISO, MIL-STD, etc.), when it comes to such a detailed level as a motion segment (MS), standardization and the definition of a 'normal' IVD is far from being a straight forward task.

Even following the traditional 'cervical/thoracic/lumbar' general division, if it is difficult to admit a resemblance between C1-C2 and C7-T1, it is certainly most arguable to sustain a similarity between L1-L2 and L5-S1, for instance. Such geometric specificities are obviously connected with the biomechanical functioning and loadings developed at each level by each individual. Besides, a quick survey can show that many FEA and MBS computer models, either in 2D or 3D, use published geometrical parameters taken from experimental data, such as those of [Natarajan et al., 1994], among others.

However, when facing computer modeling as a tool to obtain an overall 'picture' on a certain subject, and not an answer to a specific case, some kind of 'averaged' data must be used to build a single 'standard' model.

Nevertheless, a throughout sensitive analysis must also be performed, in order to access (and, if possible, quantify) the influence of main geometrical parameters, such as sagittal wedge angle, transverse length-to-width ratio or coronal nucleus-annulus relative position, not to mention other specific characteristics like the nucleus swelling capacity, that is known to vary with its size and position in the spine [Adams and Roughley, 2010].

### **2.3 Tissues' mechanical properties**

The IVD consists basically on an anisotropic solid porous matrix filled with a fluid. Thus, one may sustain that IVD numerical modeling can only be accurate enough if viscoelasticity, poroelasticity and fiber anisotropy are taken into account [Schmidt et al., 2013]. Besides, fibers resist only to traction and aqueous solutions only to compression, so an osmotic swelling pressure should also be considered, as it rules the IVD pressure under loading.

*In vivo* IVD studies are quite challenging to perform and, therefore, the major part of the information about the IVD behavior still comes from *in vitro* studies. These studies are usually performed *ex vivo*, although several recent works describe bioreactors able to keeping an excised IVD alive up to three weeks, which opens a window for new and more accurate data to be available.

## **3. Hyper-visco-poro-elasticity**

### **3.1 Modeling on the AF**

The NP is responsible for sustaining the axial compressive loads, while tensile stresses are held by the fibers of the AF, which are naturally in a pre-stressed state [Iatridis et al., 1997]. The annulus is a complex 3D concentric structure (and anisotropic), where the collagen fibers in each lamella have a criss-cross angle of about  $\pm 30^\circ$  [Shankar et al., 2009]. Mechanical performance is achieved by an ingenious combination: an external compression develops a NP internal hydrostatic pressure which is partially supported by the confining effect of the AF, whose fibers are loaded in tension, in such a way that tissues answering only in tension effectively do withstand the compression loads applied to the IVD.

For modeling purposes, mechanical properties of AF fibers can be accessed through reliable published experimental data [Holzapfel et al., 2005]. However, viscoelastic behavior of the AF relies on the characteristics of the matrix material, as clearly show the results of [Iatridis, 1999], but which can only be accessed indirectly.

### **3.2 Viscoelasticity versus poroelasticity**

Stress and strain fields within an IVD are very time-dependent. Creep and stress-relaxation phenomena have been exhaustively described in the literature, as the IVD (mainly the NP and the AF matrixes) has a rate-dependent behavior, grounded not only on poroelastic effects, but also on hyperviscoelasticity [Schroeder et al., 2010]. Maxwell, Kelvin-Voigt or Zener rheological models have been widely applied, being the generalized Maxwell model described as the best fit for IVD modeling [Ehlers et al., 2009]. Nonetheless, while hyperviscoelasticity is more important to model fast events (a few seconds), poroelasticity is far more important to describe long-term phenomena like creep and/or stress relaxation, i.e., to model the long-term behavior (minutes to hours).

### **3.3 Biphasic or multi-phasic formulation**

Most of the poroelastic IVD FEA studies are based on a biphasic formulation, which only considers the influence of solid matrix and fluid parts. While triphasic and quadriphasic theories include the influence of the ionic fluxes, in a biphasic model that flux is considered to be infinitely fast and, thus, may be dismissed.

This is an acceptable simplification leading to a smaller number of constitutive parameters and a lower model complexity. In addition, biphasic approaches showed good compromise between results accuracy and computational time, in comparison with more complex models.

### **3.4 Osmolarity**

The swelling phenomenon is due to an osmosis-driven positive pressure gradient, which drives the fluid inflow in the porous solid matrix until the saturation of the extrafibrillar space and the equilibrium.

This mechanism is responsible for the continuous activation of the fibers of the AF, by inducing a minimum and permanent tensile stress state that equilibrates the NP osmotic swelling pressure. Thus, the healthy IVD presents a permanent positive internal pressure, that ensures the integrity and responsiveness of the system, mostly due to NP's swelling properties and which is more noticeable on the disk height recovery during rest periods [Schroeder et al., 2010].

## **4. Other issues**

### **4.1 Force/motion boundary conditions**

Day by day, on an infinite diversity of environments and activities, the spine is being subjected to different kinds of loads, from slow to very fast movements, from compressive to shear. The dominant feature for the load-bearing capacity is the maintenance of the NP internal pressure and the associated AF bulging effect (confining phenomenon).

Under shear loads, it was found that the healthy IVD transfers load peripherally through the AF. Unfortunately, there is a lack of studies concerning the biomechanics of load transfer in shear or combined loadings [Bazrgari et al., 2009].

Moreover, natural body kinematics is normally driven by displacements and not by forces (i.e., to target an object somewhere – floor, shelf, etc.), and the human body shall move in such a way that objects can be reached. That is why, when some instability and/or disk degeneration occurs, the daily human activities naturally tend to overload other systems, what can trigger future problems.

#### **4.2 Natural ageing and/or degeneration**

IVD composition and geometry changes significantly during development, growth, ageing and degeneration and, therefore, changes the IVD biomechanical behavior [Adams et al., 2010].

In a normal IVD, the nucleus develops an internal hydrostatic pressure due to compressive loadings and AF confining effect. When the disc degenerates, the loading bearing mechanism changes, because the water retained by the NP diminishes and thus the tensile stresses in AF fibers are lost, as they are not “activated” by the internal NP pressure.

As unloaded/unstretched fibers tend to shrink and get stiffer with time, that means that any attempt to simulate such degenerative events have to be able to understand and model how to model/take into account the evolutive mechanical properties with time and the biophysics of degeneration mechanisms and aging. This is, currently, one of the major open questions.

#### **4.3 Role of variable ‘time’**

The spine is mechanically loaded all day long, even during rest or sleep. The response of each MS is clearly influenced by the IVD behavior and by the interaction with the surrounding musculoskeletal system, namely ligaments and muscles. In addition, the distribution and transfer of loads is dependent on the type of solicitation.

Published *in vivo* experimental data show that not only complex load sets may be produced (which imply a composition of loads and torques, acting simultaneously in the Cartesian axes, even for a simple daily activity) but also, and most remarkably, that loading rates can reach considerable high values (typically ranging from 200 to 800 N/sec.).

Thus, the real question is if plain uniaxial, quasi-static, loading conditions – as usually employed in the ‘virtual tests’ – are suitable for performance predictions under working conditions; or, in other words, if the viscoelastic behavior can be neglected.

On the other hand, the outcomes of displacement, pressure and volume variation allow the evaluation of the biomechanical behavior of the IVD osmo-poro-visco-elastic model, under different load cases of uniaxial unconfined compression. The activity load was described as three or four times averagely larger than the recovery load, i.e., it should have an average magnitude of ~700N, while moderate daily activities can represent a ~1000N load. However, the recovery process of the IVD is impaired for loads higher than 850N, which means that higher loads will probably accelerate IVD degeneration.

### **5. Applications – Ongoing Results**

#### **5.1 A lumbar IVD viscoelastic FEA model**

For accessing time-dependent kinematics of the non-degenerated IVD, under pre-defined 3D axial loading conditions (i.e. compression, flexion and torsion) a dedicated FEA model has been developed.

It incorporates a standardized geometry, the anatomically-based circumferential layout of AF fibers – from ventral to dorsal regions and their evolutive mechanical properties, both in circumferential and radial directions – and also the osmo-hyper-poro-visco-elastic properties of the lumbar MS, coupled with its poroelastic biphasic osmotic swelling behavior.

In the process of a preliminary validation against experimental published data, results seem to be in a very good agreement, both in terms of compressive force and intradiscal pressure. However, for the time being, discrepancies on the prediction of extension motion showed the need of new improvements (mesh geometry, introduction of major external ligaments,...).

Nevertheless, results already performed for creep assessment were inside the scope of the experimental data, with a remarkable improvement of the numerical accuracy, when compared with previously published results.

The main objective of this model is to be used in the prediction of mechanical performance alterations, due to modifications in nucleus physical constitution. Simultaneously, characteristic curves of response to uniaxial and multiaxial stress states, drawn from this model, are intended to be used as an input to a virtual model of the lumbar spine using a different simulation approach.

#### **5.2 Built up of a multibody system (MBS) of the all lumbar spinal section**

A multibody system, which takes in consideration the ‘solid’ body masses and their connecting constraints, is a particularly proficient tool to predict dynamic motion responses of a given set to external stimulations, and/or to analyze the results of a given motion, in terms of displacement, trajectories, forces and torques internally developed.

Data obtained by the FEA model can be translated into MBS spatial constraints, representing the mechanical characteristics of the IVD – which, according to a sensitive analysis, can be identical or differentiated at each level – and the non-linear connections equivalent to all relevant ligaments in the MS can also be incorporated.

Such a model of the lumbar spine is expected to reliably predict the load spectrum at each one of the IVDs, for given movements reproducing daily activities, which in turn can be feedback into the FEA model of the IVD, for a detailed analysis of its behavior under ‘real’ conditions.

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## Acronyms

ANSI - American National Standards Institute  
ASTM - American Society for Testing and Materials (Standard Practice for Human Engineering Design)  
HFES - Human Factors and Ergonomics Society  
ISO - International Organization for Standardization  
MIL-STD - USA Military Standards (for Systems, Equipment and Facilities)

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